Preferred walking speeds and metabolic costs while using a walking-assistive walker-like exoskeleton

THESIS

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Abstract:

Humans have a sophisticated neuro-muscular system which helps them walk in a smooth and well-balanced manner. Movement disorders are mostly caused by abnormalities in central and peripheral nervous systems. Such disorders cause recurrent involuntary motions or an inability to move despite the usage of adequate strength. Better understanding of walking biomechanics has the potential to inform development of assistive devices that can aid humans with movement disorders, a way to walk better. The devices used today for physical therapy and rehab have not been able to explain how the human body interacts with them as they are complex to model and have additional variables to compute. Moreover, these devices often fall short in providing the metabolic reduction that it should produce. Hence, a simplified version of these complex setups is necessary to get a better understanding of how humans interact in detail with an external powered device. This study used a simplified Zimmerman's powered walker-like exoskeleton cart that was built at Ohio State's Movement Lab to focus on how external assistance from such an exoskeleton reduces the metabolic energy for walking and to characterize the human-exoskeleton coupled dynamics. To begin with, the prototype was improved for better data acquisitions to conduct two types of trials: (1) measure the preferred overground walking speeds for different fixed motor assistance levels and (2) measure metabolic cost on a treadmill-exoskeleton setup for different fixed motor assistance levels. We concluded that the device can substantially increase over ground walking speeds. Using a simple mathematical model, we showed that the device may decrease the energy cost of walking but only up to a certain assistance level, beyond which it would increase the cost. This

pursuit can also give us a better insight of the neuro-muscular interactions with an exoskeleton and help improve current assistive walking devices to aid people with gait disorders.

Acknowledgement:

I would like to thank my research advisor Prof. Manoj Srinivasan for believing in me and giving me this opportunity. Prof. Srinivasan encouraged new ideas and helped me build my interest in the field of biomechanics. He supported me throughout my research and was always available to guide me in the right direction. He has always been passionate about this field of research and I feel honored to have the opportunity to work alongside him.

Secondly, I would like to thank my team member, Rashid Mattar, for helping me get the exoskeleton setup ready for conducting trials. I really appreciate all those lectures on electrical systems of the cart and for all the stay backs when we faced complications. I will always remember your analogy of voltage, current and resistance.

Lastly, I would like to thank all friends and family who morally supported me and volunteered to conduct trials. They were always there to cheer me up and to create a positive environment for me to focus on my academic pursuits.

Vita

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Chapter 1: Introduction

1.1 Motivation:

Humans use walking in their daily lives to get from one place to another. They have a sophisticated neuro-muscular system which helps them walk in a smooth and well-balanced pattern. This gait pattern is largely dependent on the person's age, mood, personality, and other sociocultural aspects; for instance, people living in cities walk significantly faster than people in rural areas (Pirker et al, 2017). Gait disorders increase with increasing age: in 60-80% of adults above 80 years and in 30% of adults above 60 years (Mahlknecht et al, 2013). Old age and degenerative diseases of the nervous system are some of the main causes of gait related disorders today. Such movement disorders cause recurrent involuntary motions or an inability to move despite the usage of adequate strength (Shahzad et al, 2016).

Over the years, the fields of biomechanics and robotics have helped develop assistive walking devices and exoskeletons which have proved to reduce the physical burden required for walking by providing power-assisted locomotion and by reducing the metabolic energy necessary (Lenzi et al, 2016). However, we have a very limited understanding of how humans interact with these devices. On the one hand, we have sophisticated mechanical exoskeletons which provide leveled power assistance and control but modelling its human-device interface is complex. Moreover, they are fixed to a treadmill and restrict the user's motion (Zhang et al, 2017). On the other hand, we have walking assistive devices like walkers and rollators which provide a simple model but do not reduce metabolic cost (Zimmerman, 2016). These devices often fall short in providing the metabolic reduction that it should produce. Hence, it is necessary to set up a simple exoskeleton model to better understand the assistance provided by these devices.

1.2 Literature review:

People suffering from gait disorders require the use of mobility aids to walk. Physicians use a combination of simple assistive devices and exoskeletons to reduce the effort required for walking and to strengthen required muscles. Simple assistive devices like walkers and rollators provide stability by reducing ground reaction forces and are meant to reduce metabolic costs. However, conducted trials on a rollator show inconsistent findings of reduction in metabolics but factors like arm position adopted while using the device played a role in reduction (Hill et al, 2012). Moreover, the inability to optimize the device adds on to its limitations. Exoskeletons on the other hand, provide leveled assistance and reduced metabolics but limit the users motion to a fixed trajectory on the treadmill setup. Moreover, it is difficult to create a model to understand their human-device interaction. Powered cart-like walkers are a better alternative to these limitations. They are simple to model, portable, does not restrict the user and can be used on real terrain (Zimmerman, 2016). However, there is very little research that quantifies their reduction in metabolic costs. This study uses a simple powered cart-like walker to test the reduction in metabolic costs at different assistance levels.

Wearable assistive devices have proved to provide mobility aid for people with injuries. One major challenge in designing these devices is understanding the musculoskeletal segments that require the needed assistance. Experimental results along with musculoskeletal simulations suggest that assistance should be provided to those muscle groups that contribute more to human gait mechanism than others; devices provided to the hip or knee provide a 4.5% greater reduction in metabolic costs when compared to ankle devices (Dembia, 2017). The net metabolic reduction while using a hip assist device was 14.71% with an increase in reduction after training and gait enhancement. Subjects also improved metabolic reduction after getting familiarized with the device (Lee et al, 2019). The experimental methodology in this research uses a similar hip-based gait belt that attaches the subject to the cart for maximum assistance efficiency.

Step frequency or steps/min is a major parameter that could be used to characterize intensity of gait in humans (Wang et al, 2013). It can be calculated by knowing the time taken and the number of steps taken to get from one point to another. Studies show that a higher step frequency consumes more metabolic energy, and it increases with an increase in walking speed. At a fixed walking speed, women show a higher step frequency than men (Wang et al, 2013). Old age and gait disorders moderate a person's step pattern, by lowering the step length and by increasing step frequency (Cavagna et al, 1986). This study will also monitor the subjects step frequency and step length to better characterize their gait pattern and to find the relation between the provided assistance levels and walking intensity.

1.3 Sloan Zimmerman's work:

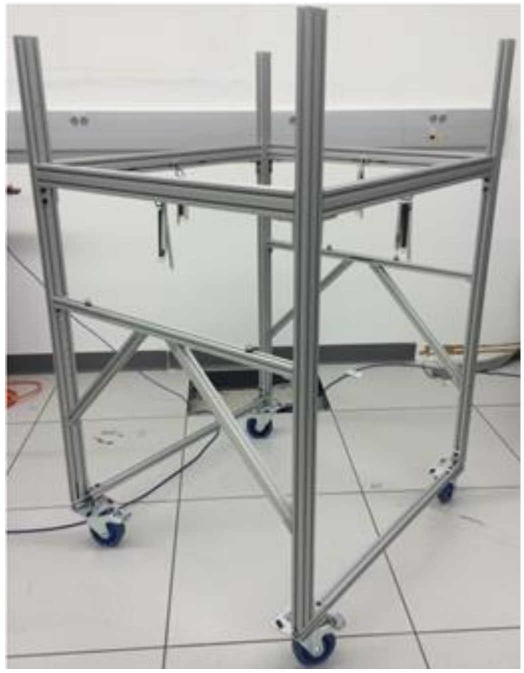

Figure 1: Zimmerman's passive walker exoskeleton assembly (without hub motor) (Zimmerman, 2016)

Zimmerman was the first to create a cart/walker-like exoskeleton (Figure 1), built at the Ohio State's Movement Lab, that simplified complex exoskeletons. This setup included a cart-like device that surrounded the human subject and provided the user with an assistive force through a set of soft springs. Zimmerman tested the effect of spring stiffness on metabolic reduction by using a range of 221 - 732N/m. Preliminary experiments determined that there was a reduction of 9% in metabolic costs while using the simulated active cart (passive cart on a treadmill set at a given speed) when compared to normal walking (without the passive cart). (Zimmerman, 2016) Zimmerman developed a simple mathematical model of a human walking while being pulled by a cart which was used to confirm experimental results.

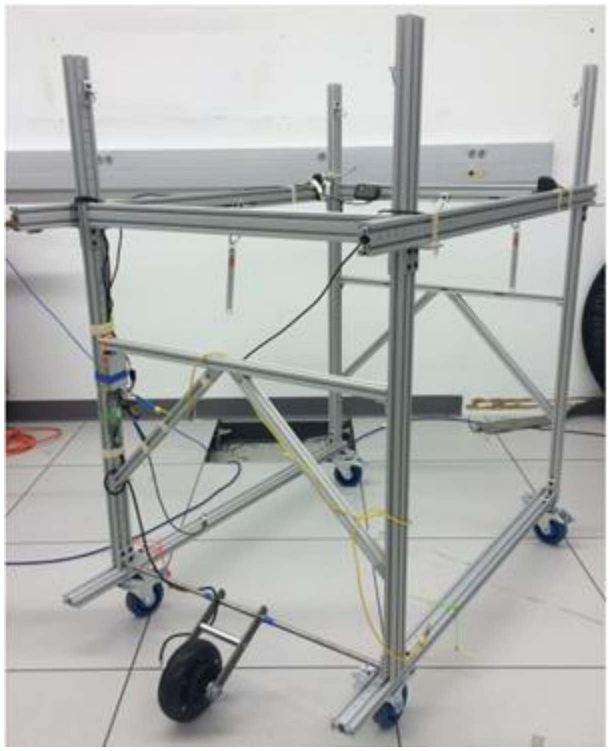

Figure 2: Zimmerman’s Powered-Cart assembly (with hub motor)

A hub motor version of the cart (Figure 2) was also tested with subjects experiencing a forward propulsive force when using the motorized walker on a treadmill (Zimmerman, 2016). However, there is still no data on the hub-motor powered version of the device and how much metabolic reduction could be achieved by this device. This research builds on Zimmerman’s hub motor powered cart to quantify the assistance provided and reduction in metabolic costs.

1.4 Goals:

The work will build on earlier work done by Zimmerman (2016). The objectives are as follows:

- To improve Zimmerman’s walker exoskeleton for better data collection and for testing our hypothesis of reduction in metabolic costs during assisted locomotion.
- To modify a throttle to provide a fixed motor assistance level that could be controlled by the user.
- To measure the preferred overground walking speeds for different fixed motor assistance levels and to find optimum user assistance.
- To measure metabolic costs on a treadmill-exoskeleton setup for different fixed motor assistance levels.
- To use the data collected to refine Zimmerman’s mathematical model of humans using an assistive device

Chapter 2: Methodology:

2.1 Background:

This study uses Zimmerman's powered cart assembly (Figure 2) to conduct the two types of experiments. The cart design had the following specifications:

Table 1: Detailed Cart Specifications (Zimmerman, 2016)

Size	80/20 framing system with T slotted extruded aluminum segments
Weight	35 kg (Light weight)
Design specifications	Accommodates height ranges from 5'0" to 6'6"
	Able to turn around in tight spaces

Figure 3: Subject attached to the cart using gait belts and springs (Zimmerman, 2016)

Attaching the cart to the subject reduced the energy required to handle the cart, hence reducing the net metabolic energy (Zimmerman, 2016). The cart was attached to the hip of the subject with the use of gait belts and springs (Figure 3). The gait belt tightly secured the subject to the cart and provided variable waist sizes. Loops were used to attach the belt to the springs. These loops were placed at the front and back of the subject as the cart provided just forward propulsive force. Threaded connectors attached the springs to the cart on one end and the belt on the other, they prevented the spring from releasing during motion. The spring chosen had a stiffness of 753 N/m as it previously provided maximum reduction in metabolic costs in this previous study (Zimmerman, 2016). Springs acted as power transmission by producing a stabilizing effect on the user and by providing reaction forces between the cart and the user that could reduce the energy required for walking.

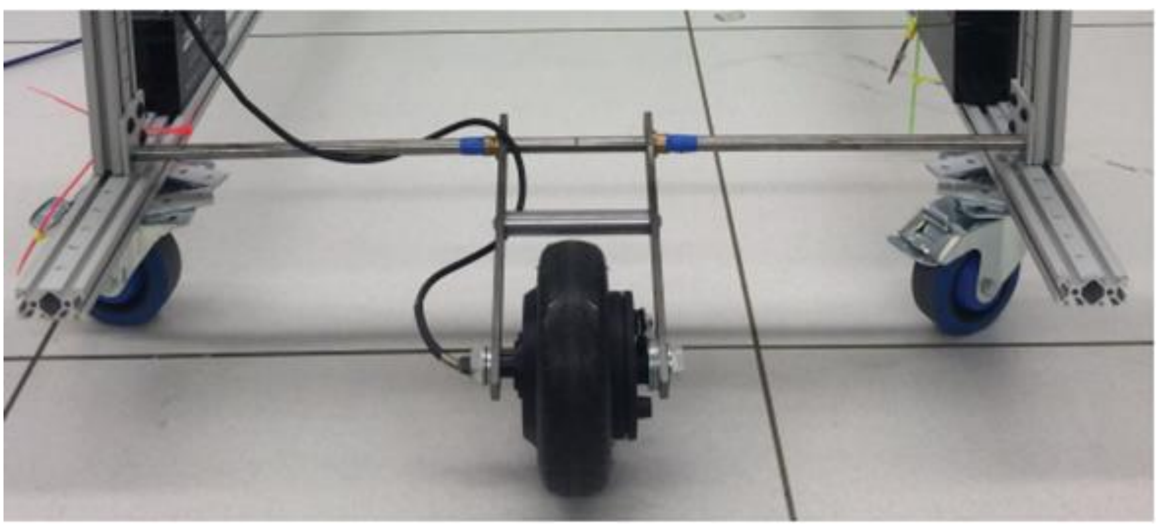

Figure 4: Closeup of hub-motor assembly

A 350 W DC brushless motor (Figure 4) with a wheel diameter of 5 inch was used to provide a forward propulsive force to the frame. The motor was attached to the front of the frame in a push configuration and powered using the provided motor controller (Zimmerman, 2016).

Figure 5: Assistance levels provided using a throttle mechanism

To conduct over-ground walking speed trials, a hall effect sensor was used to provide a set of fixed assistance levels. The sensors produced variable voltages based on the input position of the throttle. These voltages are transmitted to the motor controller which produces a proportional motor output. Considering that the parts used in this cart assembly were from a scooter setup, the small total displacement of the throttle limited the number of division/assistance levels to 5. The assistance levels started from zero coinciding with the throttles start position and ended with four coinciding with max throttle (Figure

5). Assistance level zero is the assistance provided by just holding the cart without pressing the throttle lever. The motor rpm at each assistance level was measured using a tachometer provided by Ohio State MAE's Electrical lab (Table 2).

Table 2: Translational velocity of the wheel at each fixed motor assistance level

Assistance Level	Rotational Velocity (rpm)	Translational Velocity (m/s)
0	0	0
1	60.96	0.4054
2	506.02	3.3649
3	777.8	5.1721
4	992	6.5965

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2.2 Preferred over-ground walking speed:

Our research can be divided into two parts, collecting data which involves human subject trials of healthy participants and analyzing the data to come up with a simple mathematical model. Before the start of human trials, each subject provided informed consent via a consent form. We recruited 5 healthy adults with no movement disorders. The Ohio State University's Institutional Review Board has approved this experimental protocol.

This experiment studies the change in over-ground walking speeds while using a powered cart-like exoskeleton. We will measure what speed people walk naturally without the cart, and then what speed people walk at given exoskeleton assistance. We will do that for different motor power levels (five different assistance levels). Subjects were first asked to walk naturally (without the cart) in a hallway to get a base value of their normal walking speed. This no cart trial was conducted twice to get an average value. Subjects were then asked to play with the cart and try its different levels to choose a preferred assistance. Once familiarized with the cart, a randomized order of 0-4 was chosen to conduct each assisted trial. For better observation of trends, subjects were asked to walk thrice at each assistance level, for a total of 15 walking trials with the cart.

Walking speeds were measured over a span of 30 meters and were calculated based on the time it took to cover this distance (Speed = Distance/Time). All walking trials were video recorded to measure the time taken between the start and end point. To initialize subjects to the cart, they were asked to walk a distance of about 50 m forwards along a hallway and turn 180°; this can be considered as the start point from which the subjects started walking. Subjects were then asked to use the first 20 m to fix their assistance levels and to ensure the motor has accelerated to its desired translational velocity. The last 30 m till our end point was used to take our timed trials. The video recording was also used to analyze the change in number of steps taken per second.

Chapter 3: Results

3.1 Overground walking speed data:

The mean overground walking speeds generally increased from normal walking (no cart) to increasing assistance provided by the cart with a mean percentage increase of 48% when compared to normal walking speeds. Subjects showed an increase in walking speeds while using assistance level two, three and four. A decrease in walking speeds was seen for assistance zero and one. However, only subjects whose normal walking speeds were above 1.4 m/s (average walking speeds) showed this decrease while using assistance level one. Most subject trials had similar walking speeds while using assistance level zero (Figure 6).

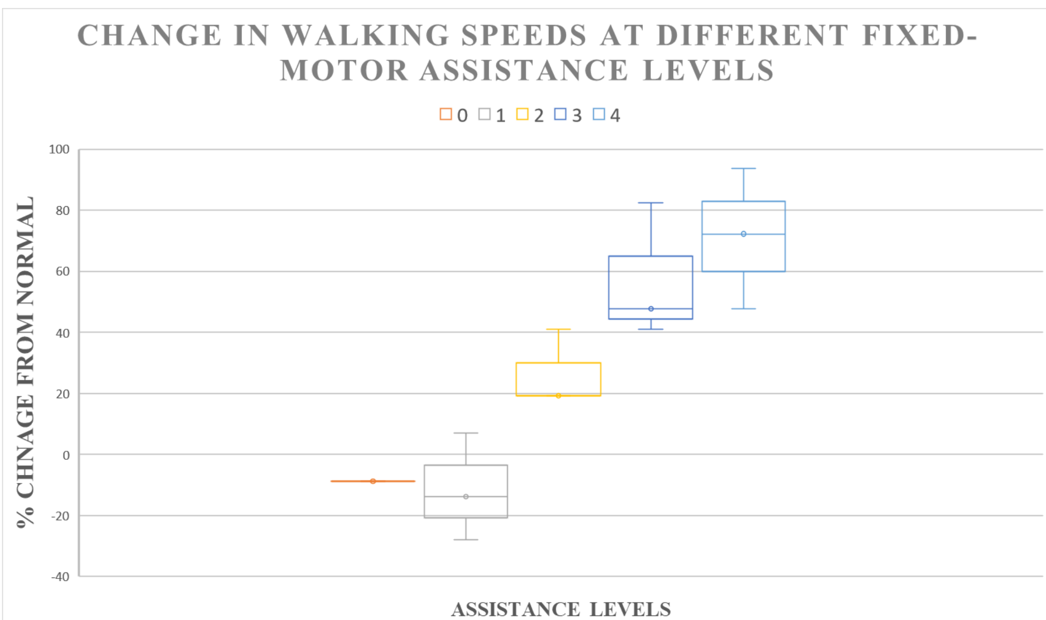

Figure 6: Change in overground walking speeds when compared to normal walking speeds (without cart) at different fixed motor assistance levels provided by the hub-motor powered cart

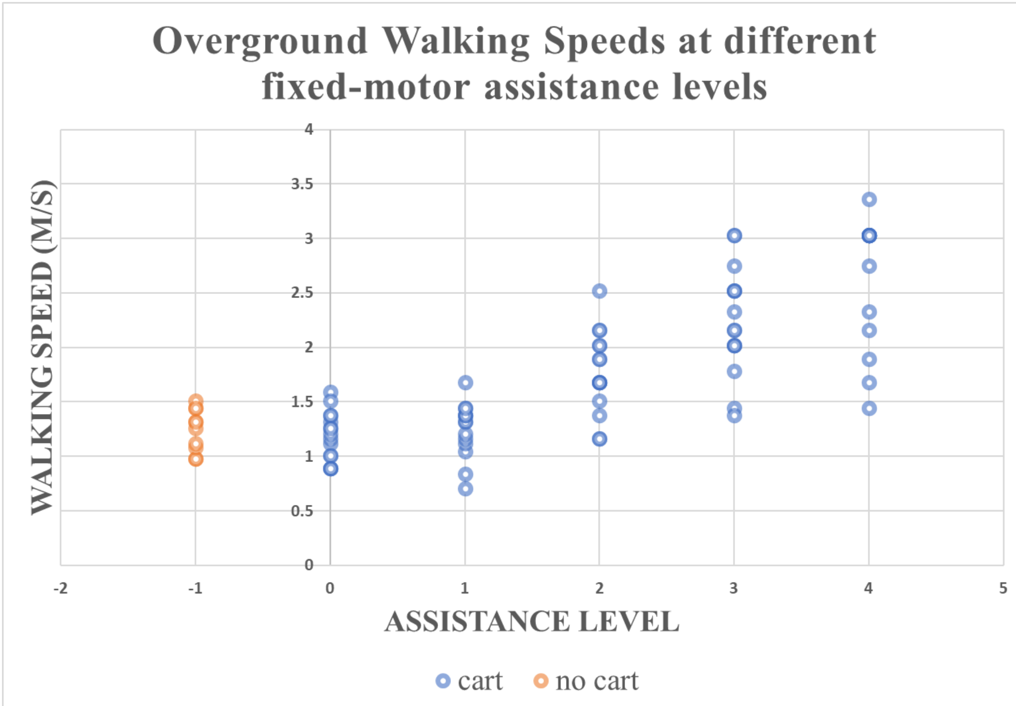

Figure 7: Comparison of overground walking speeds of subjects for no cart and cart assisted trials. Negative assistance of -1 is used as a placeholder for no cart trials

Subjects who had a normal walking speed above 1.4 m/s preferred assistance level two and felt that assistance level one provided negligible assistance and that assistance level three and four provided too much assistance. On the other hand, subjects who had a normal walking speed below 1.4 m/s preferred assistance level one, with assistance level two, three and four to have provided too much assistance. Considering that the average speed where humans equally walk and run is 2.2 m/s (Long et al, 2013), most subjects were seen running while using assistance levels three and four. Further, recall that assistance levels 2, 3 and 4 correspond to free-spinning wheel speeds of 3.3, 5.1, and 6.6 m/s (Table 2), which are much faster than even jogging speeds. Nevertheless, with the wheel attached to the cart and the cart attached to the human, the overall speed of human + cart was, respectively, around 1.9, 2.2, 2.5 m/s on average (Figures 6-7). These ‘compromise’ speeds are obtained because the cart is dragged back by the human and human is being pulled forward by the cart.

3.2 Metabolic costs derived from mathematical model:

In this approach, Zimmerman's telescoping biped model of humans walking (Zimmerman, 2016) was used to determine the metabolic costs at each assistance level for different overground walking speeds. Here the spring stiffness (k) of the spring connecting the cart and the human is used to model the forward propulsive force or assistive force provided by the cart. Assuming a human of mass (m) 70 kgs, leg length (L) of 1 m and acceleration due to gravity (g) is 10 m/s^2, Real stiffness = Normalized Stiffness *(m*g)/L = Normalized Stiffness *700 N/m. As per the chosen geometry, the cart allows a maximum deflection (x) of about 8 cm (Zimmerman, 2016). The forward propulsive force (F) can then be modelled as k*x N. Six spring stiffness in the range of 0 and 700 N/m were chosen to provide a forward propulsive force up to 56 N to build on our analysis (Table 3) . Here spring stiffness zero is used to model a no cart setup.

Table 3: Relation between spring stiffness and forward propulsive force

Normalized Spring Stiffness	Real Spring Stiffness (N/m)	Forward Propulsive Force (N)
0	0	0
0.2	140	11.2
0.3	210	16.8
0.5	350	28
0.8	560	44.8
1	700	56

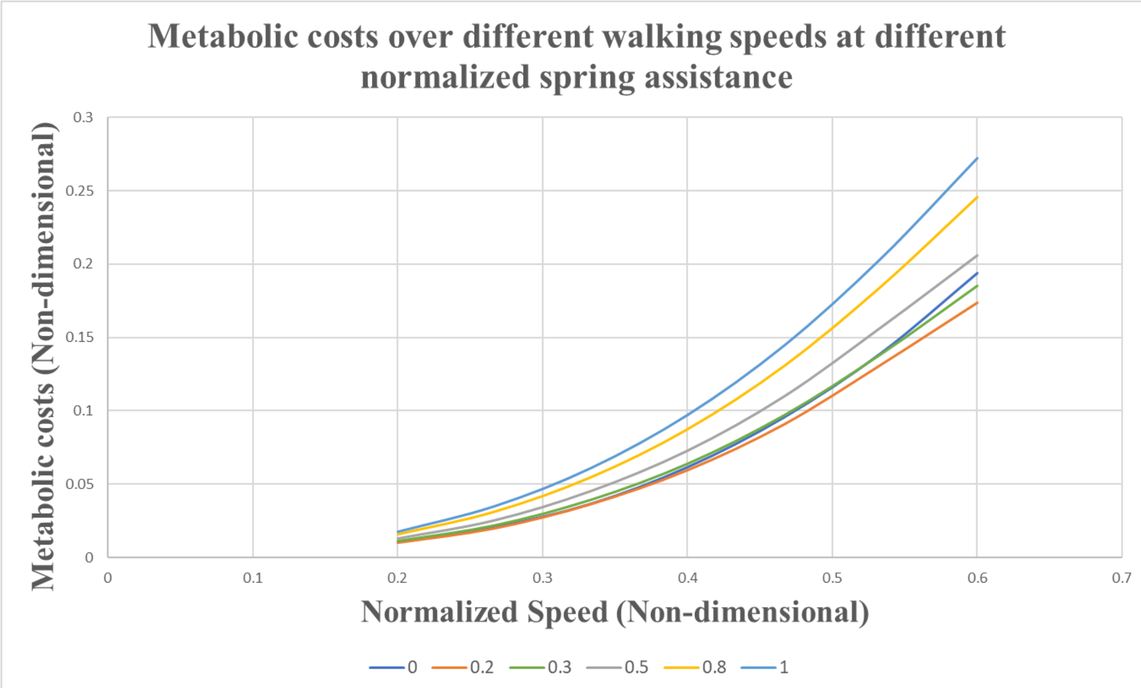

Figure 8: Metabolic energy needed to maintain overground walking speeds for different spring assistance levels

The above model (Figure 8) shows an increase in metabolic costs per unit time (metabolic rates) with an increase in overground walking speeds. This is true for all walking, whether assisted or unassisted. The metabolic rates also increase when much more assistance is provided with a mean percentage increase from no cart to be about 22% across all the conditions we considered. However, a reduction in metabolic costs is seen for different walking speeds when the normalized spring assistance is below 0.3 (F = 16.8 N). The mean reduction for normalized spring stiffness 0.2 was 5%. Hence, the optimum forward assistance provided by the exoskeleton should be between 0 and 16.8 N for metabolic rate reduction.

3.3 Step frequency and Step length data:

In Figures 9 and 10, a general increase in both step frequency and step length are seen as assistance levels are increased. However, at lower assistance levels (assistance level one), there is an increase in step frequency and a decrease in step length as subjects found it easier to maintain the cart's trajectory. At higher assistance levels, a non-uniform gait pattern was observed with subjects regularly increasing or decreasing their over ground walking speeds as they found it difficult to maintain the cart's trajectory.

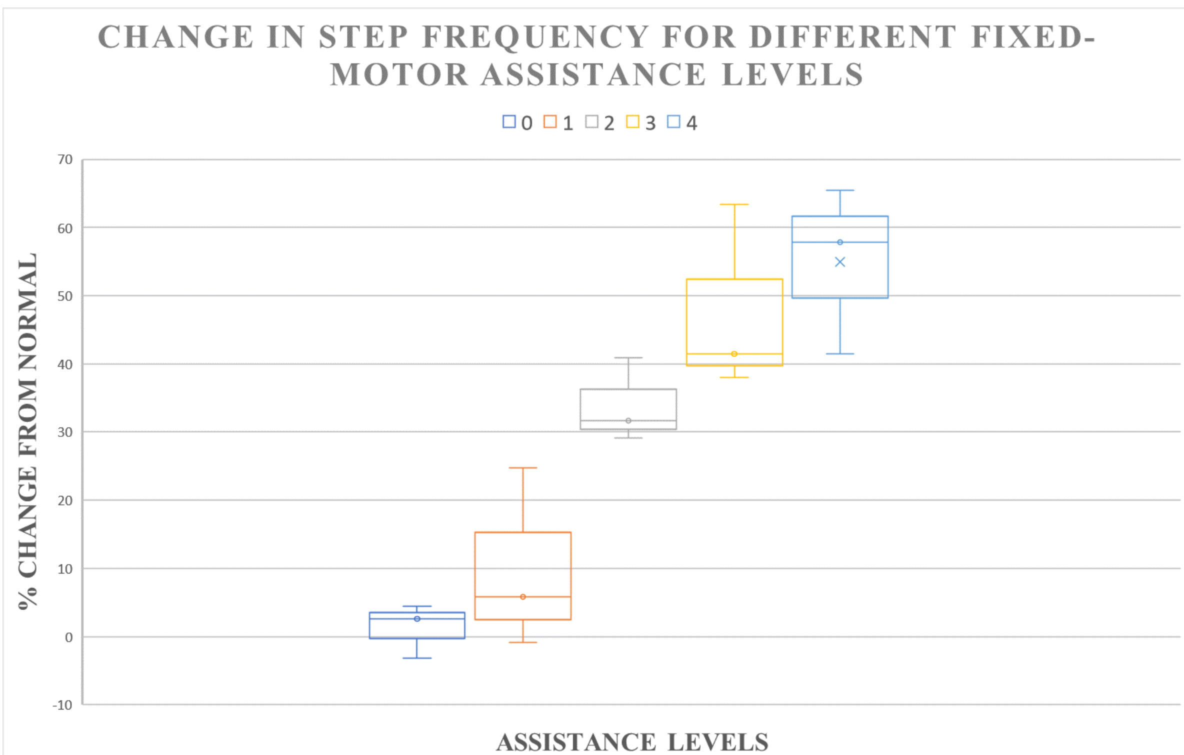

Figure 9: Change in Gait Intensity (steps/sec) for different fixed motor assistance levels when compared to normal walking (no assistance)

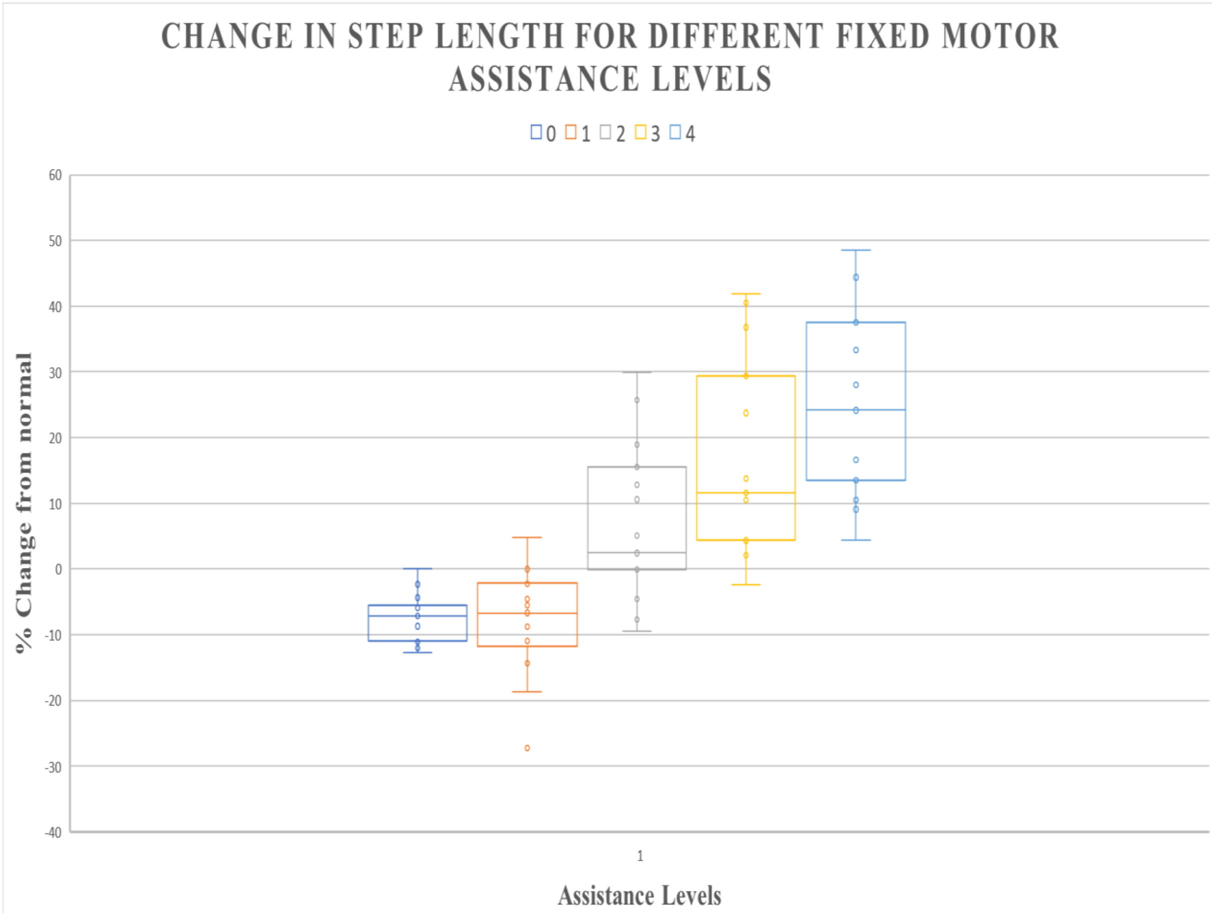

Figure 10: Change in Step Length for different fixed motor assistance levels when compared to normal walking (no assistance)

Chapter 4: Discussion

4.1 Over-ground walking speed analysis:

As seen in Figure 6, subjects in the zero assistance level trials showed a decrease in overground walking speeds. A reduction proves that additional energy is required to drag the cart along the direction of motion. Subjects whose normal walking speeds were above average (1.4 m/s) showed a decrease in overground walking speeds while using assistance level one and felt that it did not provide any assistance. However, an increase in walking speeds was seen at their preferred assistance level two (Figure 6). This means that subjects tried to walk along with the cart rather than drag it along with them. Hence subjects had to walk at a lower speed to maintain the assisted speed provided by the cart. On the other hand, subjects whose normal walking speeds were below average preferred assistance level one and showed an increase in overground walking speeds at their preferred assistance (Figure 6). The evidence provided can be used to hypothesize that subjects with higher walking speeds require higher assistance to maintain their speed. In addition, as gait disorders also reduce overground walking speeds to below average, one can say that the cart is successfully able to increase overground walking speeds with a minimum assistance like assistance level one.

4.2 Analysis of metabolic costs derived from mathematical model:

As seen in Figure 8, which describes metabolic effects using a model, a reduction in metabolic costs can be seen only till normalized spring assistance level of 0.3. This means that an optimum assistance level exists above which the assistance can be considered as “too much”. Too much assistance will also make the subject walk at a faster speed when compared to normal walking speeds and hence the subject will have to exert more metabolic energy to maintain that speed. Effects of too much assistance can also be seen in Figure 7, as subjects are seen to run rather than walk while using assistance level three and four. Moreover, subjects found it more difficult to handle the cart at higher assistance levels. Adding results of overground walking speed measurements (Figure 6) to this analysis, one can also conclude that the optimum assistance required is higher for higher normal walking speeds. Further, in the regime the metabolic costs is decreased for a given speed, the optimal walking speed increases, partly because the subject can now walk or run faster for the same cost as when un-aided.

4.3 Analysis of step frequency and step length data:

The forward propulsive force of the cart on the human is analogous to walking down a slope and gravity pulling the human down the slope. As step length is lower and step frequency is higher at lower assistance levels (Figures 9 and 10), the subject using the cart can be considered walking down a hill where one often takes smaller steps on an incline. Also, the assistance level provided can be related to the steepness of the hill. A steeper hill, which relates to more assistance, increases the walking speed while coming down and hence makes it more difficult to walk. Hence, subjects are seen running down the hill with faster and longer steps. When compared, it is evident from Figures 9 and 10 that subjects at higher assistance levels show a similar trend of increase in step frequency and step length.

Chapter 5: Conclusion and Future Work

5.1 Summary:

This research focused on improving Zimmerman's powered cart-like exoskeleton for quantifying the assistance provided and the reduction in metabolic costs by conducting overground walking speed trials for different fixed motor assistance levels. A hall effect sensor was used to provide five fixed assistance levels. The data collected was first analyzed for trends and then compared to Zimmerman's mathematical model of "humans using an assistive device". Major gait parameters like overground walking speeds, step frequency and step length were used to characterize the subject's gait pattern.

It was determined that the cart was successfully able to increase overground walking speeds for non-zero assistance levels with an average increase of 45% when compared to normal walking speeds. A reduction in walking speeds at zero assistance showed that some energy is required to drag the cart along the direction of motion. Metabolic costs were only reduced until an optimum assistance level, above or below which the assistance provided can be considered as "too much" or "too less". The optimum assistance is higher for higher normal walking speeds. With too much assistance, subjects were seen to walk faster (run) and exert more metabolic energy to maintain the added assistance. On the other hand, subjects with low assistance were seen to walk slower but exert more metabolic energy to maintain reduced speeds.

The step frequency and step length data was used to model a new assisted terrain. As humans generally take smaller steps on an incline, a decrease in step length and increase in step frequency observed at lower assistance levels can be used to conclude that the device can create a scenario of walking down a hill. Further, a higher assistance can be compared to a steeper slope as subjects were seen to run faster with an increase in both step frequency and step length.

5.2 Future Works:

1. A further analysis is needed to find the optimum assistance level needed to produce a reduction in metabolic costs. Due to time constraints, the team was unable to conduct treadmill trials with metabolic measurements. Treadmill trials will measure motion, ground reaction forces and metabolic energy at three treadmill speeds and five exoskeleton assistance levels. With these measurements we can estimate a metabolically optimal preferred walking speed and also a metabolically optimal motor assistance level at a given speed. The experimental method includes video capture and placement of reflective markers on the subject's torso for the measurement of position data, force plates for the measurement of ground reaction forces and a portable metabolic measurement device (Oxycon Mobile) for measurements of metabolic costs. See Figure 11 and 12 for lab setup.
2. Future work will also involve modifying Zimmerman's mathematical model of humans using a cart like exoskeleton. The role of the speed controller on the cart needs to be accounted for creating a more exact model of the human-device interface. How the cart slows down in response to human forces is something to be considered.

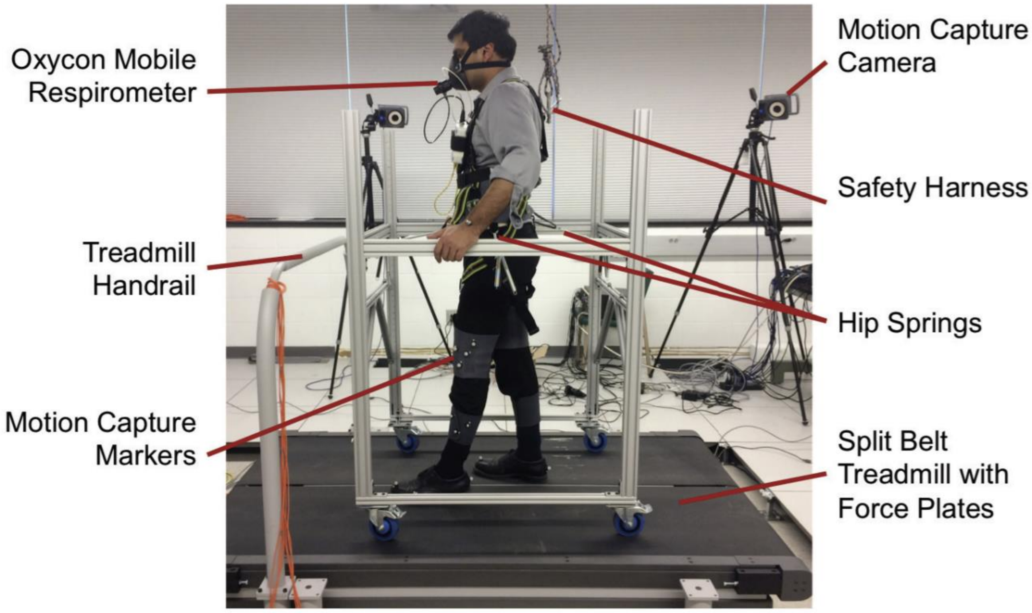

Figure 11: Passive cart treadmill trials (Zimmerman, 2016)

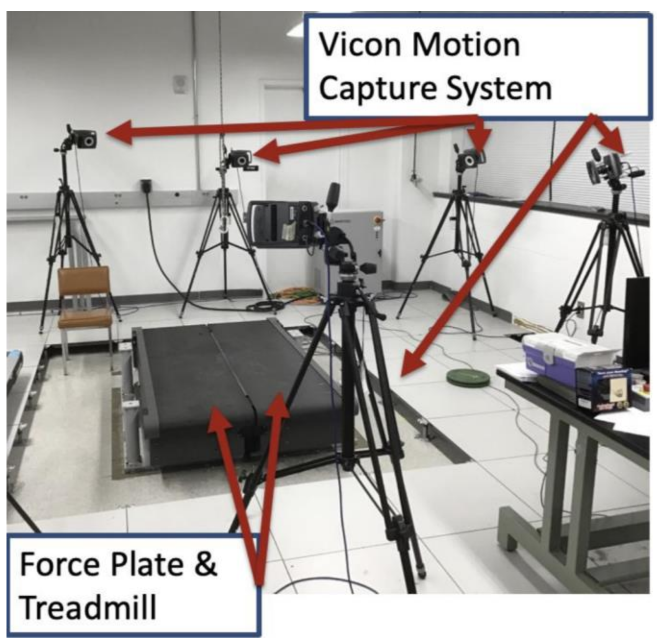

Figure 12: Experimental setup for treadmill trials at the Movement Lab, Ohio State (Zimmerman, 2016)

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